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BEHAVIOR OF MECHANICAL MODEL OF
SYSTEMIC CIRCUIT OF CARDIOVASCULAR SYSTEM

by

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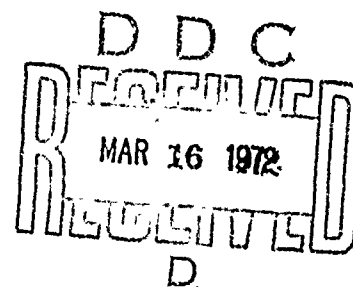


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BEHAVIOR OF MECHANICAL MODEL OF SYSTEMIC CIRCUIT OF CARDIOVASCULAR SYSTEM

ABSTRACT

A simplified physical model of the systemic circuit of the human cardiovascular system was previously designed and constructed for the purpose of studying certain mechanical aspects of fluid flow in such systems [1]¹. With the model an extensive investigation of several aspects has now been made. It consists essentially of experimental studies of the effect of variation of rigidity of the tubes, of the effect of bleeding on performance, and the effect of the variation of frequency of drive on systemic behavior. A preliminary report was previously made on the first two subjects. For these, a significant addition is described in the present report. Also, important findings concerning frequency effects are given in some detail.

¹Numbers in brackets refer to references found on page 20.

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I. INTRODUCTION

Various kinds of models and analogues are available for the study of many kinds of phenomena. A classification of types of physical models has been attempted by G. Murphy in his textbook on the subject [2]. A sophisticated model study of a complex problem of astrophysics in terms of gas dynamics is given by L. I. Sedov [3]. On account of the similarity of difficulties in our present biological problem and those in Sedov's astrophysical problem it may be worthwhile to quote him directly. He says, "It has now become evident that the formulation and solution for a number of dynamic problems of gas motion, which can be considered as theoretical models encompassing the essential peculiarities of the motion and evolution of stars and nebulae must underlie the conceptions used in the investigation of celestial phenomena. In order to construct and to investigate such models, methods, apparatus, and conceptions of modern, theoretical gas dynamics must be used, namely aerodynamics, and the appropriate mechanical problems applicable to astrophysics must be formulated and solved." The success of Sedov in his investigation should lend encouragement to anyone attempting to study complex systems with simpler mechanical models. In contrast to his work, we compare the mechanical functioning of a hydroelastic recirculatory system with that of certain aspects of the human cardiovascular system.

It seems worthwhile to emphasize that while only a single prototype exists in the case of the astrophysical modelling, there really are as many specific prototypes as there are humans in the case of the cardiovascular system. Notwithstanding this fact, the general abstract

principles of cardiovascular performance are of fundamental importance to physiologists. Although the individual examples of the systems are so astoundingly complex, and even the physiological abstractions themselves are complex, it is our considered opinion that certain features of the mechanical model can meaningfully throw light on certain corresponding probabilistic mechanical features of the prototype. In any event it is quite clear that our approach constitutes a rational starting point for so challenging and important a problem.

While it may be of interest to consider here in some detail the relationship of biological modelling to the general subject of modelling it will only be recalled that the author has treated the matter in a previous paper [4]. The present report concerns itself only with a model study of certain mechanical aspects of the systemic circuit of the cardiovascular system. The specific topics which are treated explicitly are the effects on performance of the system caused by a change in rigidity of the tubes (models of the arteries), bleeding off of liquid, and controlled change of frequency of pulsations.

II. MODEL PUMP

A so-called elastodynamic pump was previously developed for the purpose of making model studies of the human cardiovascular system. It was fully described in a previous technical report [1]. Since that time the pump has been improved and completely tested. In the original model the check valves used were rubber flap valves. These had certain operational defects and were consequently replaced by a more satisfactory design which is shown in Fig. 1. It may be noted that the flap type check valves are replaced by silastic ball check valves. These valves utilize very light guides attached to the orifices and have carefully designed coil springs for seating the silastic balls against O-rings with predetermined forces. The complete pump and driving mechanism is shown in Fig. 2. With the exception of the valves, it is much the same as the previous model. However, it should also be noted that the spring used with the cam follower has been redesigned in order to prevent break of contact between cam and follower throughout the range of operation. The original time-displacement relation is still employed as shown in Fig. 2. The systolic phase of the pulsatile motion is sinusoidal and the diastolic phase is constant and approximately of zero displacement. It may be recalled that the body of the pump consists essentially of two rubber hemispheroids attached one to each end of a short cylindrical shell to form a compact fluid container. It is readily seen that a properly designed forcing mechanism will produce a unicursal flow in any fluid system into which the pump is inserted.

An external view of our pump is contrasted with views of the standard diaphragm pump, the DeBaKey pump, and the so-called Army pump in

Fig. 3. The essential difference between our pump and each of the others is quite clear. Ours has a deformable housing which is caused to change shape by the driving mechanism. Each of the other pumps has a rigid housing but an internal elastic membrane or diaphragm.

The notion of pump characteristic experiments to determine performance readily suggests itself. Since it is well-known in technology that data from such experiments are indispensable for the application of all standard types of pumps, it is clear that they are equally well required for our pump. One distinction of these deformable elastodynamic pumps with pulsating drive is that the usual fixed pressure across the pump for a given volumetric delivery must be replaced by a time integral of the periodic pressure. Such has been readily obtained in the present study by means of diaphragm type pressure transducers inserted in the flow line. The pressures were readily recorded as functions of time on oscillographic recorders after suitable amplification. For determining the pump performance, straight tubes upstream and downstream were used. The liquid flows into the pump from a reservoir and is pumped downstream against measurable resistance to a graduated container where the volumetric flow is timed. The apparatus for these experiments is shown in Fig. 4. It may be seen that two arrangements of the experimental apparatus were used. Type (a) provides a downstream variable resistor in the form of an adjustable cone throttling device. Type (b) has a variable height reservoir or liquid container which provides a predetermined water head against which the liquid is pumped.

The performance curves for the pump are given on Fig. 5. The flow-rate Q is shown as a function of pressure head. For both arrangements (a) and (b), the pressure is measured at the pressure transducer and plotted as head in centimeters of water. Data are shown not only for the two arrangements but also for two sizes of cam. One cam provided a maximum displacement of the pump wall of one-half inch. The other cam provided a maximum displacement of one-quarter of an inch.

While all of the pump data obtained were smooth and reproducible, it can be seen that the shapes of the curves are different for the two different methods of producing downstream resistance. The curves for the case of pumping against a head of water are nearly hyperbolas, with the flowrate rising sharply with reduction of pressure head. Of course, for the case for which a cone-type resistor is used there is a definite cut-off pressure when the cone is fully closed. It should be observed that the tubing is elastic and distends with increase of pressure. How this is accomplished seems to have an effect on pump performance.

Since the pump was intended for use in experimental systems in which the tubing is distensible it was desired to use such tubing in the pump performance experiments. Of course, the same experiments can readily be conducted with rigid metal tubes replacing the latex used in our experiments.

III. MECHANICAL MODEL OF SYSTEMIC CIRCUIT OF CARDIOVASCULAR SYSTEM

In his textbook on medical physiology, Guyton gives on page 249 a simplified schematic representation of the whole cardiovascular system showing the essential parts. It can be seen that the two major circuits are the systemic and the pulmonary, which are coupled together in one complete unit. In contrast with Guyton's drawing the present report deals only with a model of the systemic circuit which was fully described in our previous report [1]. For convenience a slightly modified form is shown in Fig. 6. The pump represents the left ventricle and the latex tubing represents the arterial-venous system. It may be noted that two flow arrangements are provided. One is a continuous flow-through system of tubing subsequently referred to as a closed system. The other is an alternate arrangement which provides a reservoir open to atmosphere as shown in Fig. 7. The latter system is referred to as the open system. In all of the subsequent data presentation and discussion, these different types of models are designated as the closed and open reservoir systems.

It may be well to emphasize that our system contains no cybernetic or feedback control systems. Actual cardiovascular systems are said to have feedback controls to maintain an average reference pressure level. As described by A. C. Guyton [5], the pressure in the cardiovascular system is controlled by a feedback mechanism based on the pressure at a point in the neighborhood of the tricuspid valve of the right atrioventricular orifice. He says, "If the pressure at the tricuspid valve rises slightly above normal, then the right ventricle fills to a greater extent than usual, causing the heart to pump blood more rapidly than usual

and thereby decreasing the pressure at the tricuspid valve toward the normal mean value. On the other hand, if the pressure at this point falls, the right ventricle fails to fill adequately, its pumping decreases, and blood dams in the venous system until the tricuspid pressure again rises to a normal value. In other words the heart acts as a feedback regulator of pressure at the tricuspid valve". Such controls can be worked into a model, but it is considered that at this point in our research significant findings can be made expeditiously without using the more elaborate type.

IV. EXPERIMENTS TO DETERMINE EFFECT OF BLEEDING ON SYSTEM PERFORMANCE

In order to study the effect of traumatic conditions caused by inflicting wounds on the cardiovascular system, it was considered possible to obtain at least some phenomenological knowledge by means of simple physical models. Accordingly, a provision was made in our model to simulate the condition of bleeding. In order to perform bleeding experiments, the device shown in Fig. 8 was used. It provides a precision control for either intermittent or continuous bleeding. Intermittent bleeding means that the valve on the bleeder is opened and a predetermined quantity of liquid is drawn out of the system. The valve is then closed and the pump provides a definite flow in the model. Continuous bleeding means that the valve on the bleeder is opened and liquid is continuously run off from the model. During the latter process the rate of flow is continuously measured. For the closed system without elastic reservoir the initial pressure and volume conditions are as follows:

Tubing	Volume c.c.	Pressure above atmospheric cm. of water
Latex	2700	0
	2900	150
Vinyl	2750	0
	2790	90

For the closed system with an elastic reservoir inserted in the line the initial pressure and volume conditions are as follows:

Tubing	Volume c.c.	Pressure above atmospheric cm. of water
Latex	2800	0
	3500	60
Vinyl	2850	0
	3450	60

When bleeding experiments were conducted with the initial line pressure at atmospheric and the bleeder valve open, air intake prevented any reasonable operation of the model. Accordingly experiments were conducted with initial volumes and pressures as shown in the tables.

The effects of bleeding on the system operating with an elastic reservoir in the line are shown for both latex and vinyl tubing in Fig. 9. Also, the effects of bleeding on the system operating without an elastic reservoir inserted in the line are shown for latex and vinyl tubing in Fig. 10. In these Figures, V is the initial volume of liquid in the system and ΔV is the small volume of liquid which is run through the bleeder. The relative rigidities of the systems and of the tubings are shown in Fig. 11. In this Figure V_0 is the initial volume of liquid in tube or in system as indicated and ΔV is the change of volume of liquid. The quantity P is atmospheric pressure and ΔP is the change in pressure.

In the first phase of decrease of pressure, rate of flow decreases in all cases, but very rapidly for the case of vinyl tubing both for intermittent and continuous flow. It is considered that the radical difference between the performance of the system with latex tubing and that with vinyl tubing is at least partially explained by the very different compliances shown by the curves in Fig. 11.

It may be noted that for all of the curves in Fig. 9 and Fig. 10 the data at the largest values of the bleed off ratio $\Delta V/V$ correspond to the situation in which the internal pressure is atmospheric. It had been observed that the further reduction of pressure rapidly reduced the flowrate to zero and the system became inoperative. It is conjectured that a strictly similar situation does not exist for the actual cardiovascular system because of possible cybernetic controls and local reactions at the site of the wound.

V. EXPERIMENTS TO DETERMINE EFFECT OF TUBE RIGIDITY ON FLOW

It would seem that a change of rigidity of the tubing shown in Fig. 6 should have an effect on the performance of the model. Accordingly, for the same motion of the driving mechanism, experiments were conducted for tubing made of latex, tygon, vinyl and steel. As a consequence a very large range of rigidity was investigated.

The variation of flowrate as a function of rigidity is shown in Fig. 12. The rigidity is defined as the ratio of $\Delta P/P_0$ and $\Delta V/V_0$. The P_0 is atmospheric pressure and ΔP is the change in pressure. The V_0 is the initial volume of liquid and ΔV is the change in volume. The rigidity for the steel is so high that it is not shown in Fig. 12. The corresponding flow for that case is practically zero as might be anticipated. If an elastic reservoir were inserted into the line in the case of steel the results would obviously be different.

The experimental results are no doubt very significant in demonstrating the possible effects of tube rigidity. It must be emphasized however, that if a highly distensible elastic reservoir or a rigid reservoir open to the atmosphere were inserted into the line, the flowrate would be altered significantly. Because of this fact the results are really useful only to indicate that change of rigidity in any portion of the tubing may be of importance. In a more extensive physiological investigation the question should be more thoroughly studied.

VI. EXPERIMENTS TO DETERMINE THE RELATION OF FLOWRATE TO PULSATING FREQUENCY OF SYSTEM

Experiments were performed to determine the effect of change of pulse rate, or frequency of drive, on the flowrate in the model shown in Fig. 6. Two possible arrangements of the model were used in the experiments. In one of them, the tubing was continuous as shown. In the other, reservoir shown on Fig. 7. was inserted into the line. The former arrangement is referred to as the closed system; the latter is referred to as system with open reservoir.

Experiments were made using latex, vinyl and steel tubing. It may be noted, as pointed out previously in the report, that considerable flow will occur in the model using rigid steel tubing if an open reservoir is inserted into the line.

For the system with open reservoir, the effect of amplitude of drive was investigated for two cases. One was a maximum amplitude of pump of one-quarter of an inch; the other was one-half inch.

Experimental findings for the case of open reservoir are shown in Figs. 13, 14, and 15. Those for the case of a closed system are shown in Fig. 16. It can be seen that for all of the curves in those figures distinctive maxima and minima occur.

VII. DISCUSSION

The results of the study strengthen the principal investigator in his conviction that the use of physical models will provide useful knowledge of the functioning of biological systems.

The experimental program demonstrates the importance of the effect of tube compliance on the functioning of the model. While it may seem obvious that this is so, only carefully controlled laboratory experiments will provide precise knowledge of the fact.

The experiments performed for the purpose of demonstrating qualitatively the effect of bleeding show that the functioning of the system is substantially impaired when the internal pressure is in the neighborhood of atmospheric pressure. One of the most important consequences of a trauma to the arteries by inflicted wounds is that it will lead to the rapid induction of air into the system. A reliable analytical relation between rate of flow of blood and quantity of loss of blood is impossible of attainment from model experiments at the present time. It seems that critical conditions arise too quickly after damage is inflicted. It must be considered however, that any defensive mechanism of a cybernetic nature which may exist in the prototype are completely absent from the type of model which has been studied.

It might be stressed at this time that the elastodynamic pump, which was developed for the study, very effectively performs its function. Furthermore, extensive experience gained with the pump during the study indicate that:

- a) It should make a good heart booster (as an alternative to the DeBaKey pump).

- b) It can be used effectively for research either with physical models of the left ventricular circuit or by proper combination with a second pump it can be used for research on the entire cardiovascular system, which includes the pulmonary circuit.
- and c) It could be a useful substitute for the diaphragm type of pump which has so long been associated with special technological applications.

To our knowledge, the distinct maxima and minima in volumetric flow as a function of pulse rate, for a given thrust of the pump, is a phenomenon which has never been reported in the engineering literature on pumps in general or in the literature devoted to the functioning of the cardiovascular system.

The essential mechanical system is definitely a recirculatory one with distributed mass, flow mass, and compliances. A dynamical analysis of such systems has not as yet been made by anyone, but such would obviously be of interest. Although the liquid used in the experiments was water, there can be little doubt that the fundamental response features would remain even if more rheologically complicated liquids are used.

Textbooks on physiology do not appear to treat of the phenomenon which we have found in our elastodynamic recirculatory model. A plausible reason for this is that no one has studied the flow of blood in an animal as a function of pulse rate. The medical importance of our finding is that we have a definite suggestion that the heart delivery varies substantially with pulse rate, for a given thrust in a very definite pattern of maxima and minima. What clinical value may attach to this fact is difficult to say but it seems clear that the problem should be studied thoroughly in the human as well as in other animals.

It should be emphasized that the model which was subjected to study was always horizontally supported, thus removing any substantial variation in hydrostatic pressure. The effects of such variation could readily be studied by simply rotating the platform on which the model has been mounted from the horizontal to the vertical position. Such possible effects will be studied in the extension of the present program.

Finally, it may be stated that most of our research in the immediate future will be directed toward obtaining knowledge of the functioning of a more complete cardiovascular system, which contains both the systemic and the pulmonary sub-systems. It is only in this way that an understanding of the interaction between the left ventricular and the right ventricular pumps can be developed.

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Finally, it is a pleasure to acknowledge the very helpful discussions of some of the physiological aspects of the investigation by Dr. H. Melvin Knisely of the Medical University of Charleston, South Carolina.

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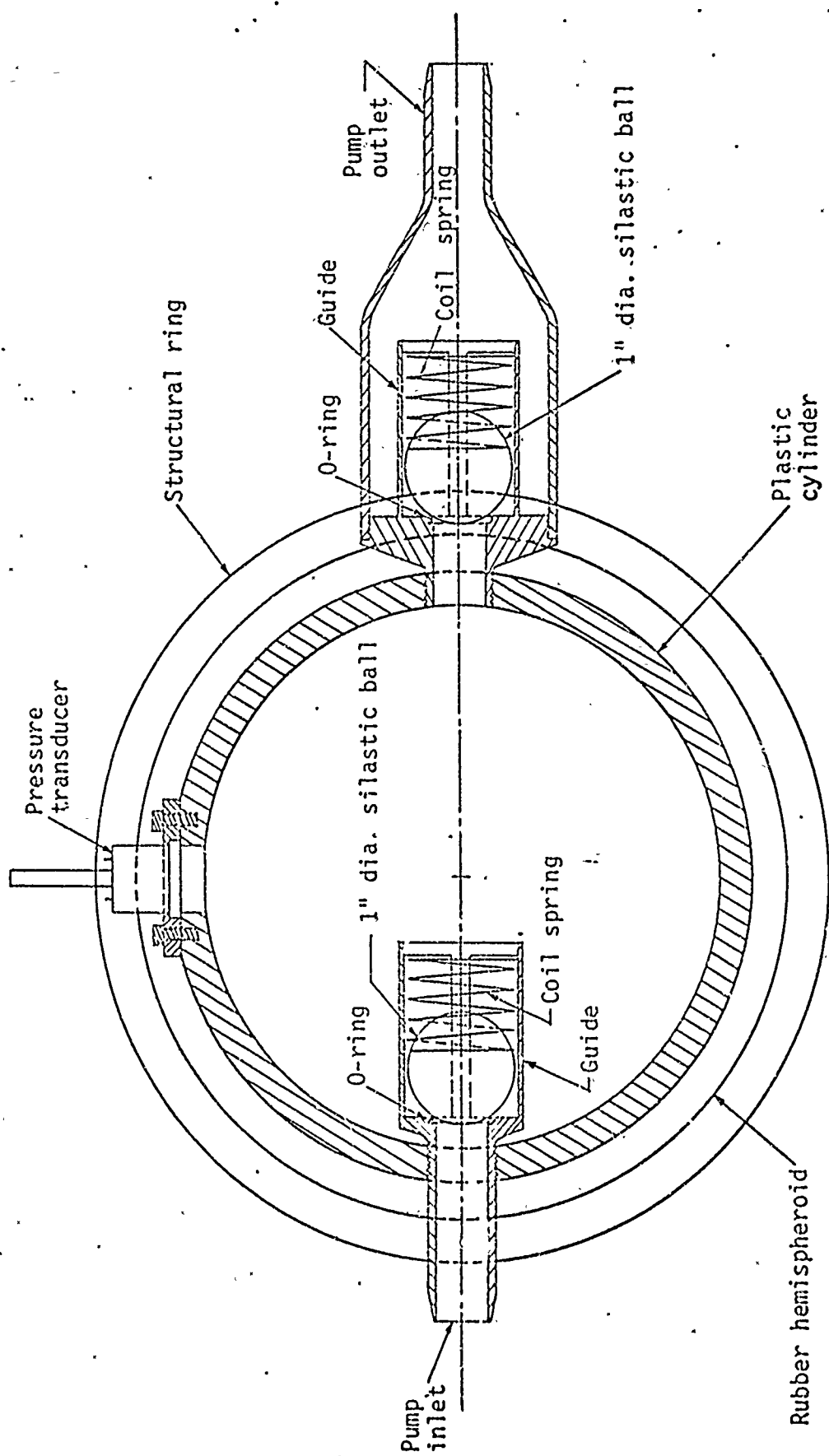


Fig. 1 Cross-sectional view of Hoppmann pump.

$$X = X_0 \sin \omega t \quad 0 \leq t \leq \frac{\pi}{\omega}$$

$$= 0 \quad \frac{\pi}{\omega} \leq t \leq \frac{2\pi}{\omega}$$

X_0 = Maximum displacement
 ω = Angular velocity of cam
 T_0 = Period of the periodic motion
 $= \frac{2\pi}{\omega}$

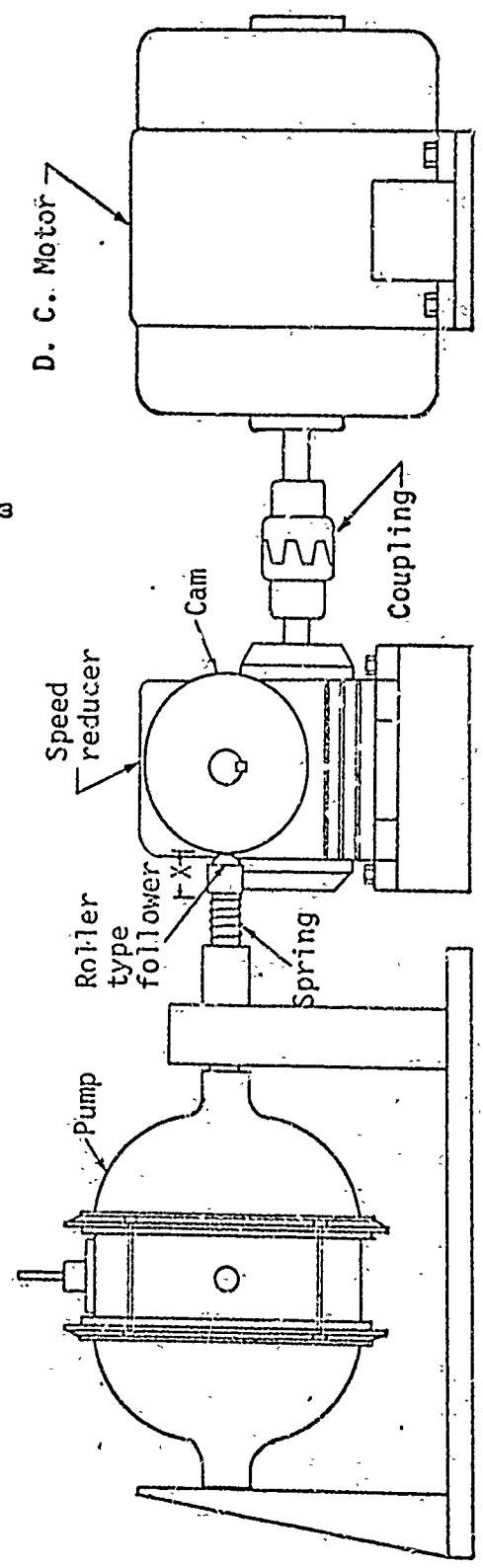
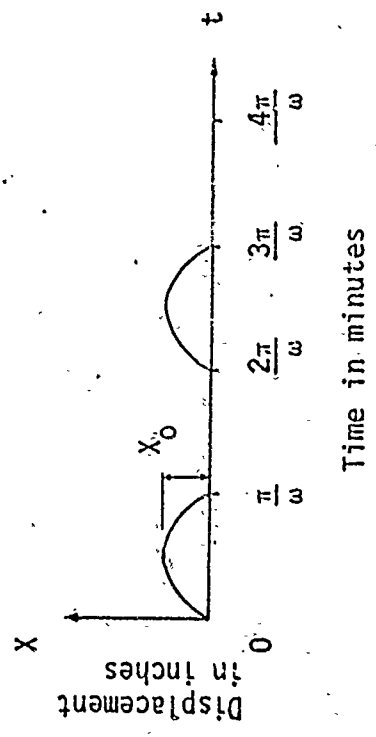
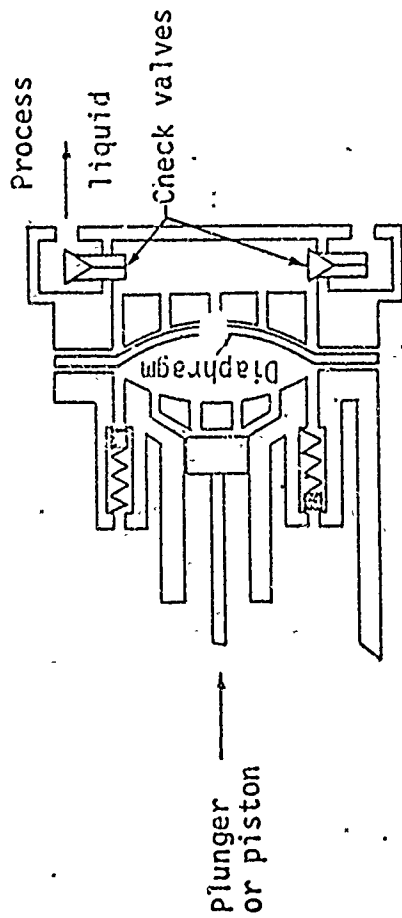
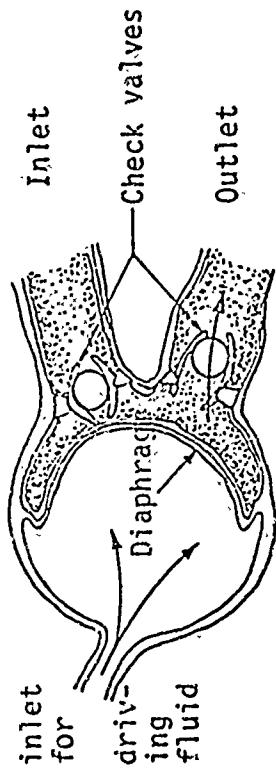


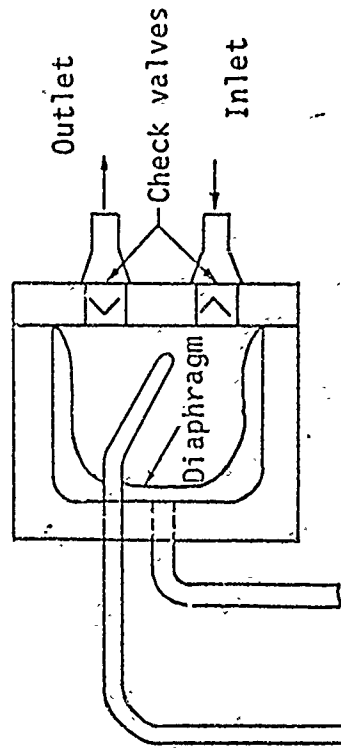
Fig. 2 Cam type driving mechanism for pump.



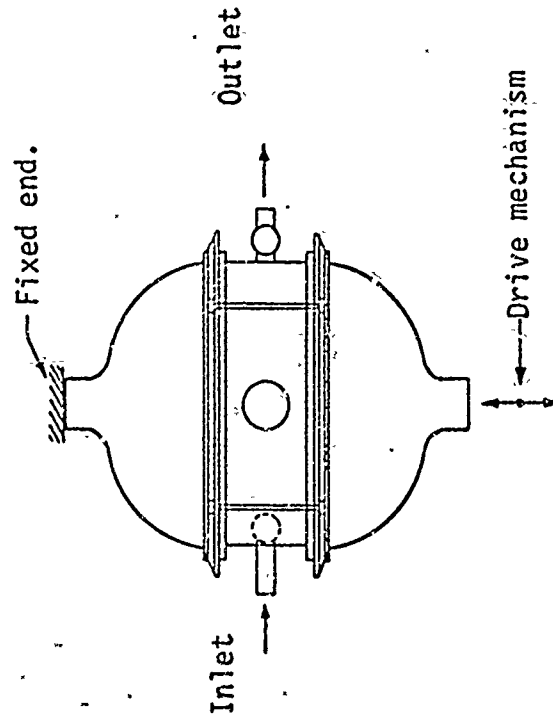
a) Diaphragm pump (discharge position)



b) de Bakey pump



c) Army pulsatile pump



d) Hoppmann pump

Fig. 3 Comparison of several types of pumps.

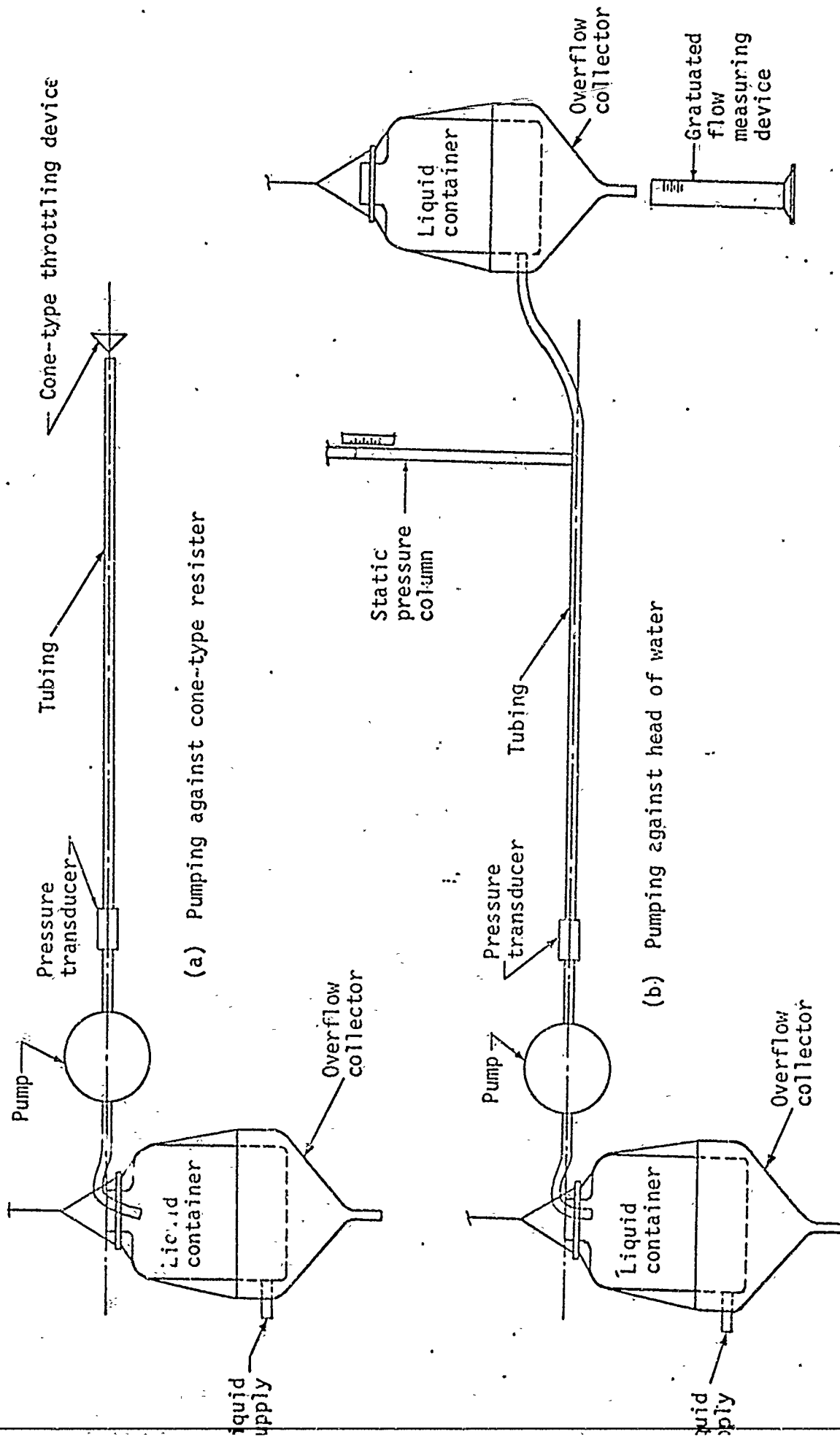


Fig. 4 Apparatus for determination of pump performance.

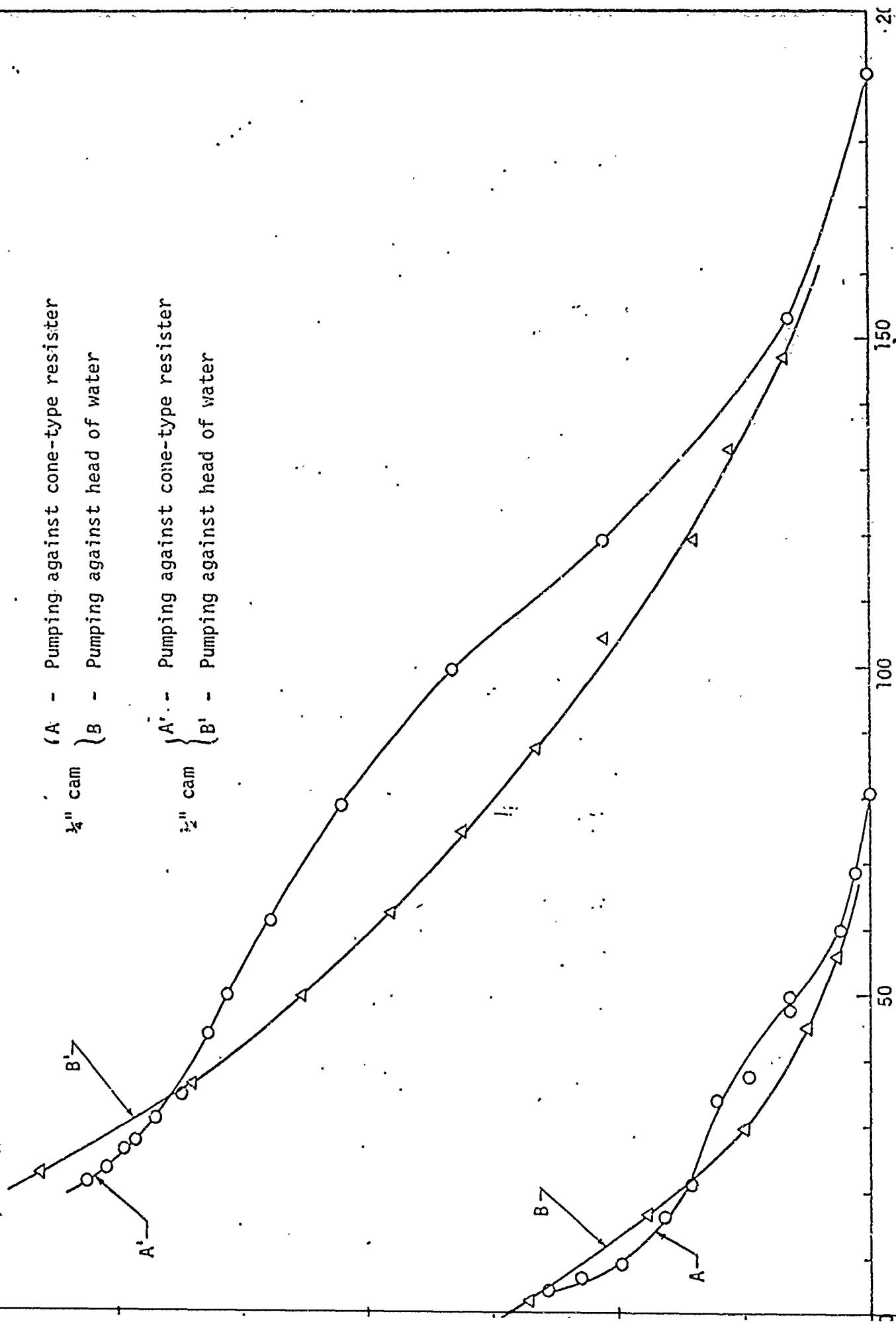


Fig. 5 Performance of pump

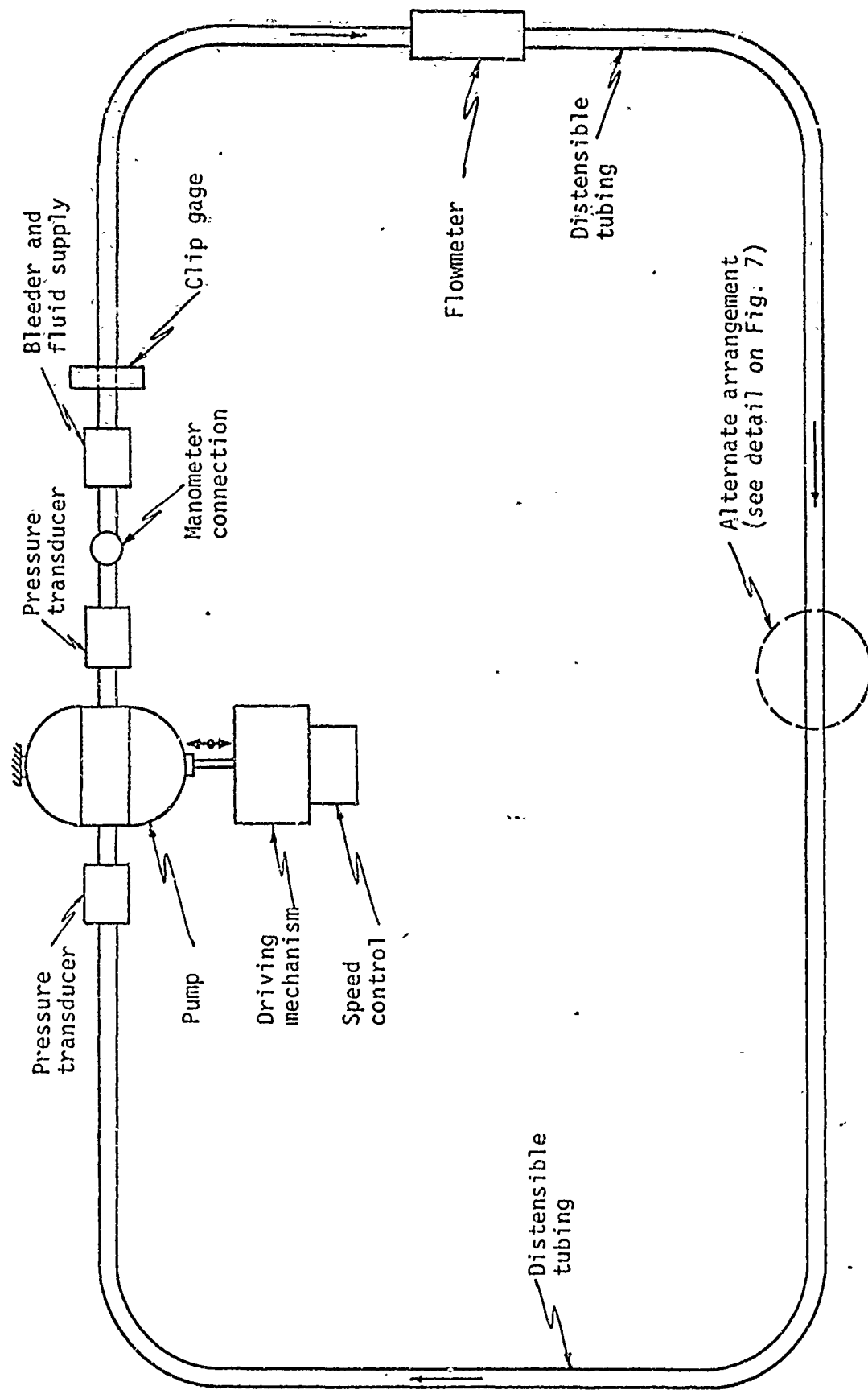


Fig. 6 Model of the systemic circuit of the cardiovascular system

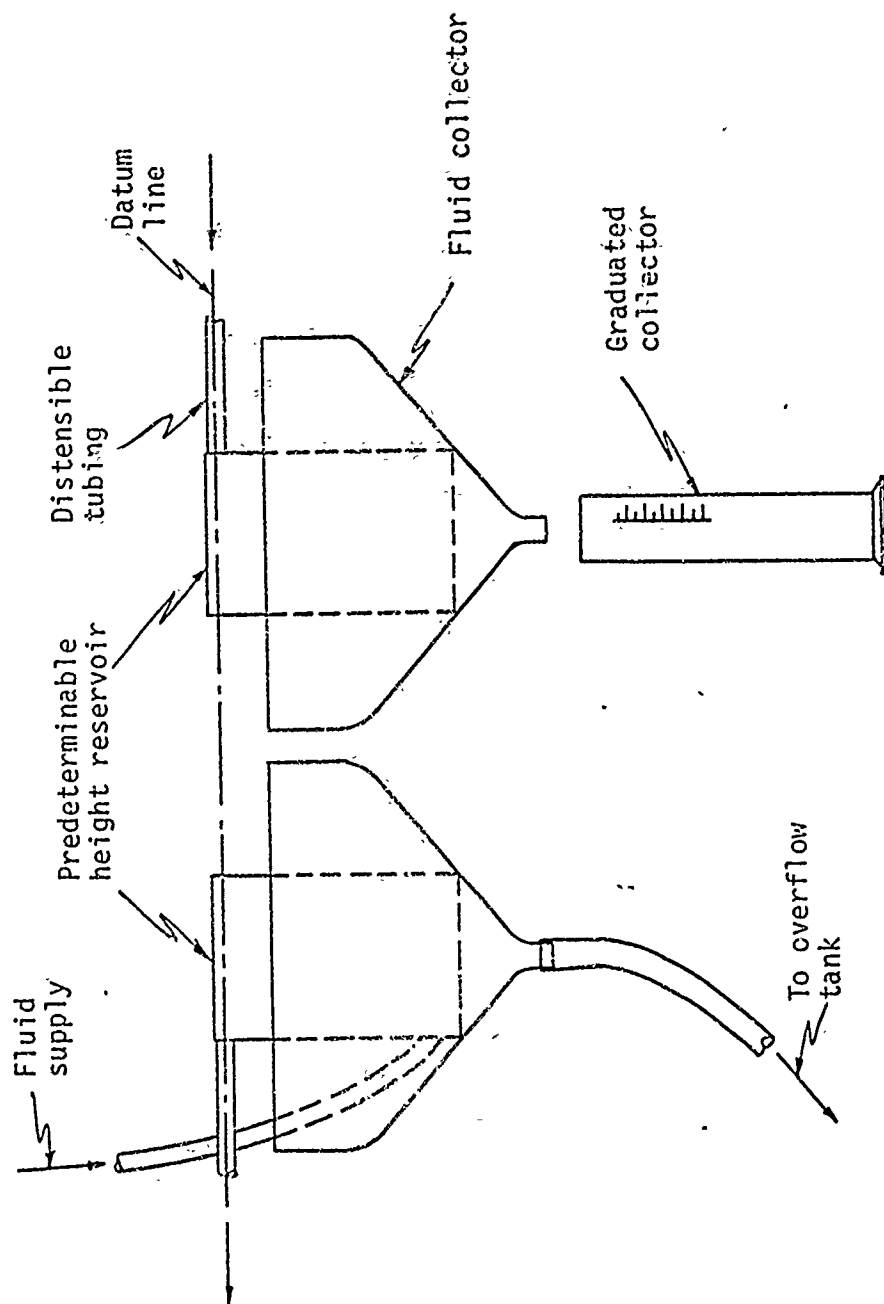
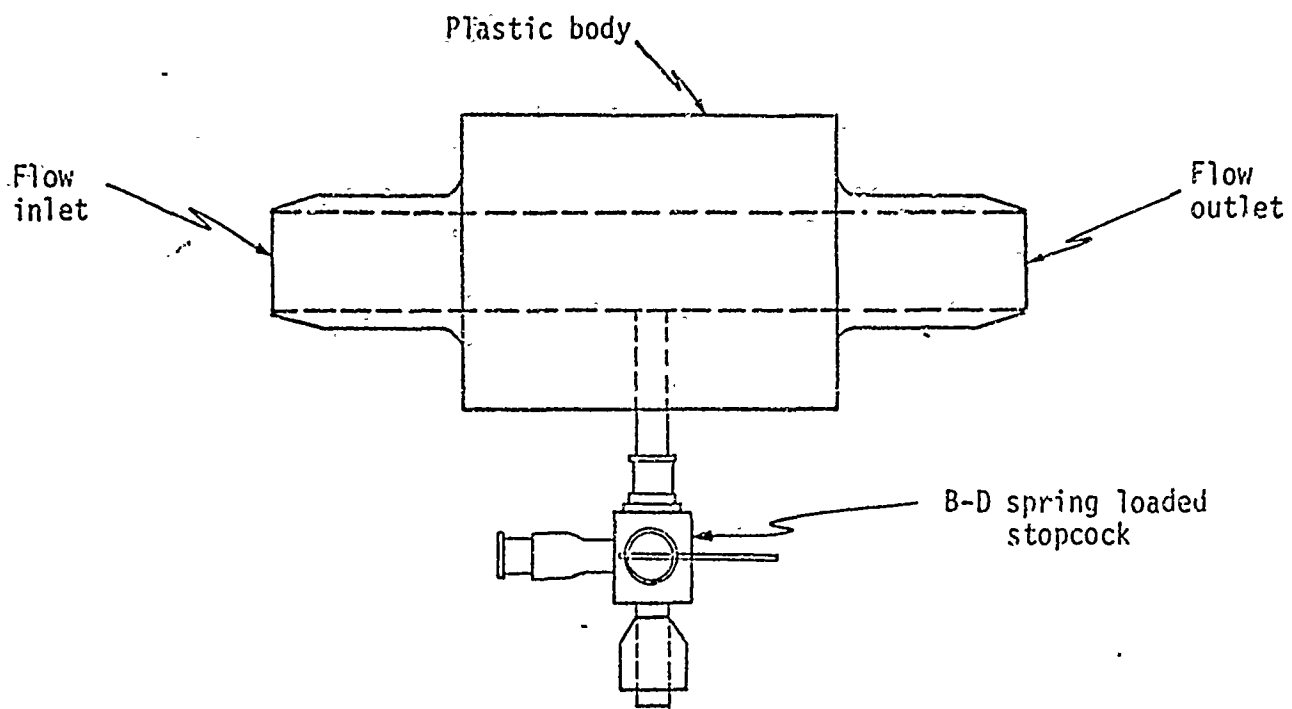
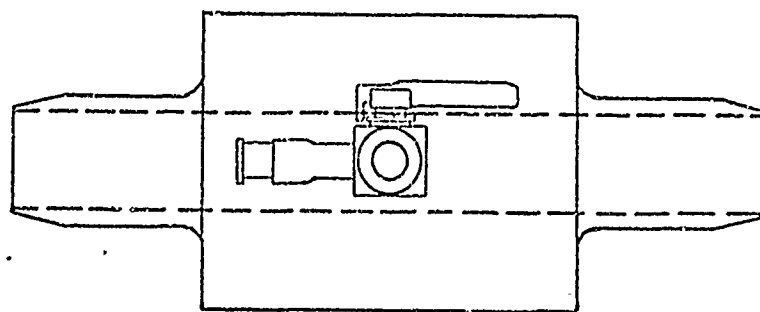


Fig. 7 Detail of alternate arrangement: open reservoir



Top view



Side elevation

Fig. 8 Detail of bleeder used on cardiovascular model.

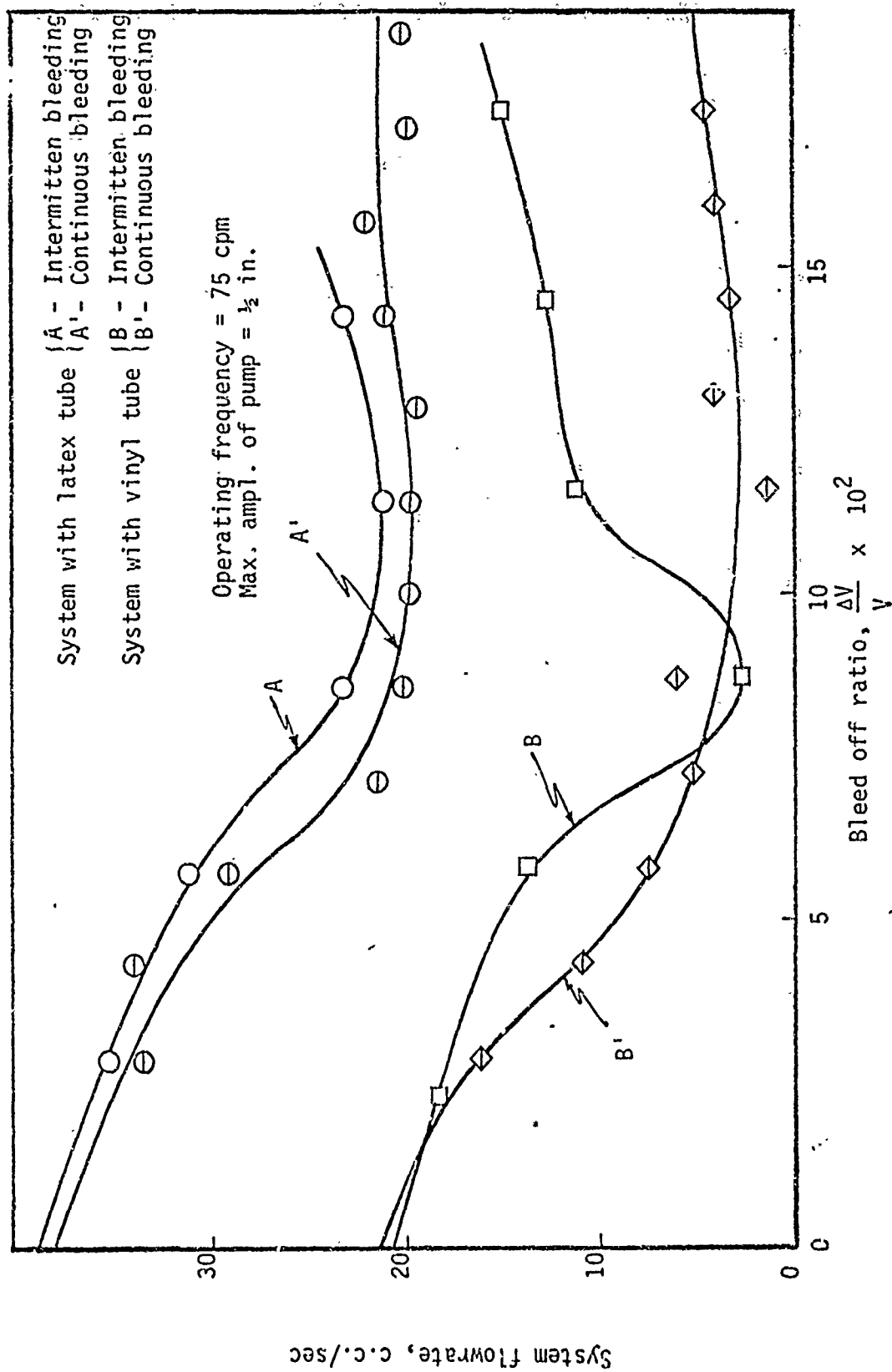


Fig. 9 Effect of bleeding on performance of system with elastic reservoir

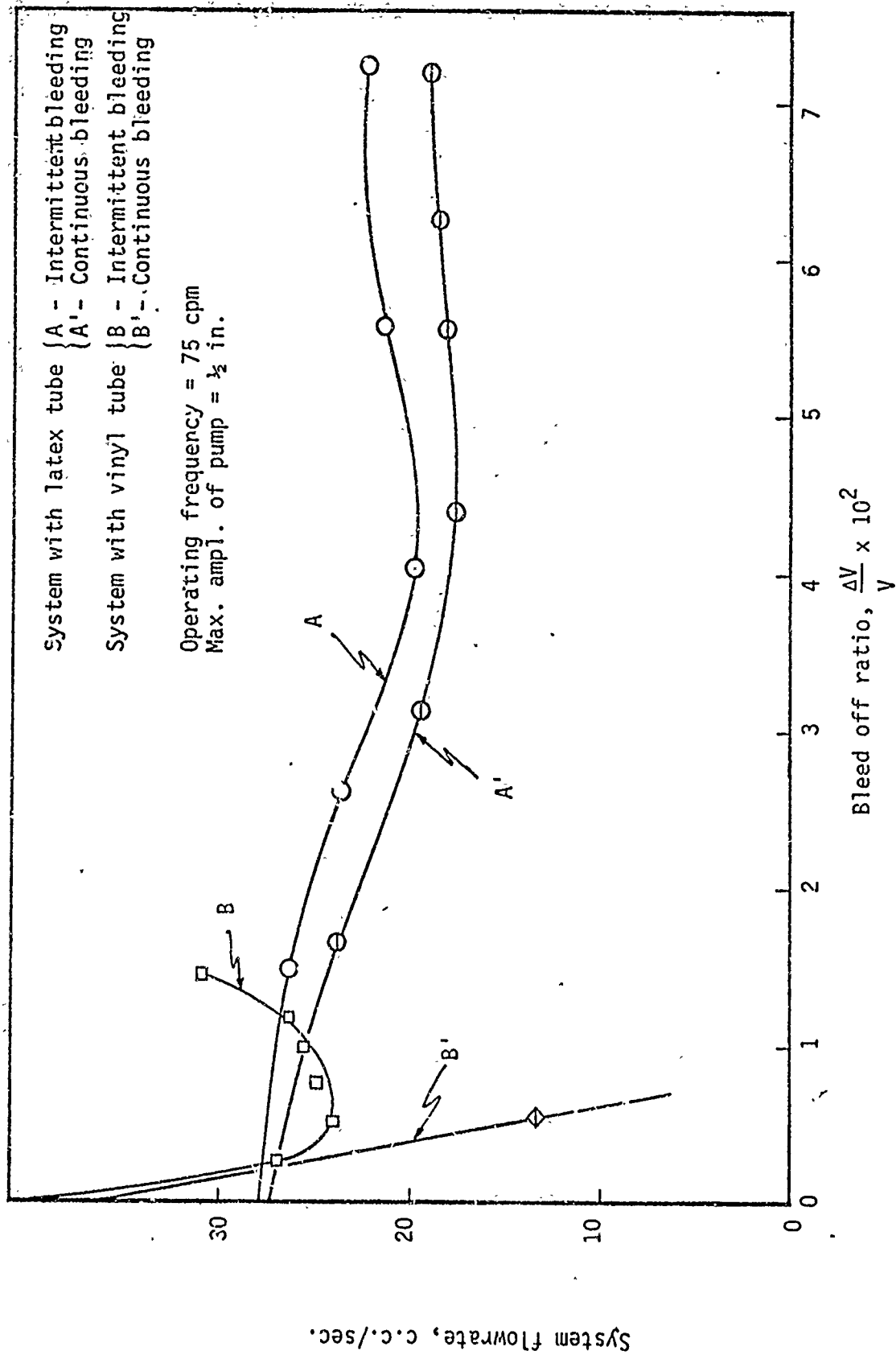


Fig. 10 Effect of bleeding on performance of system without elastic reservoir

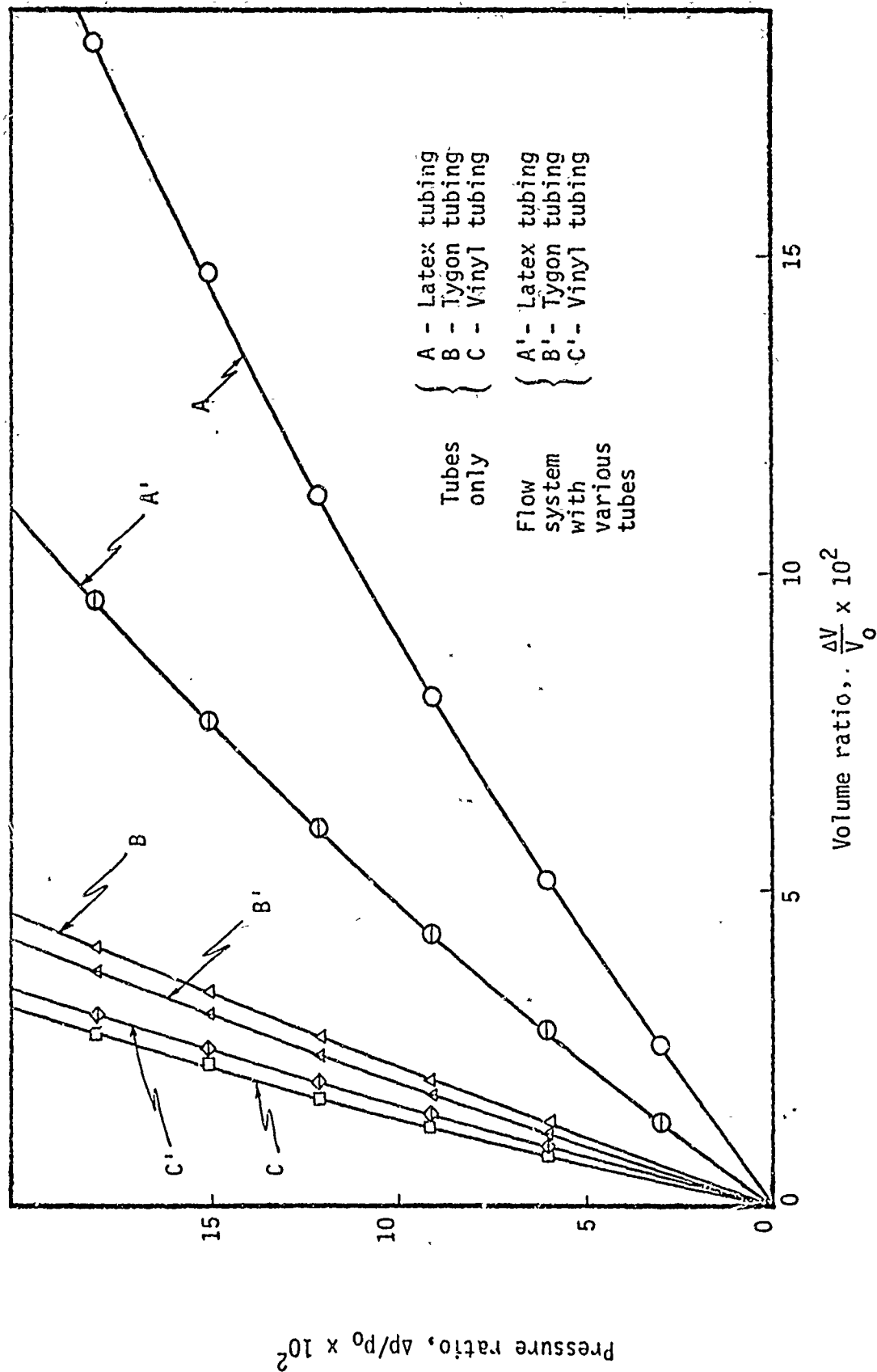


Fig. 11 Variation of pressure ratio with volume ratio

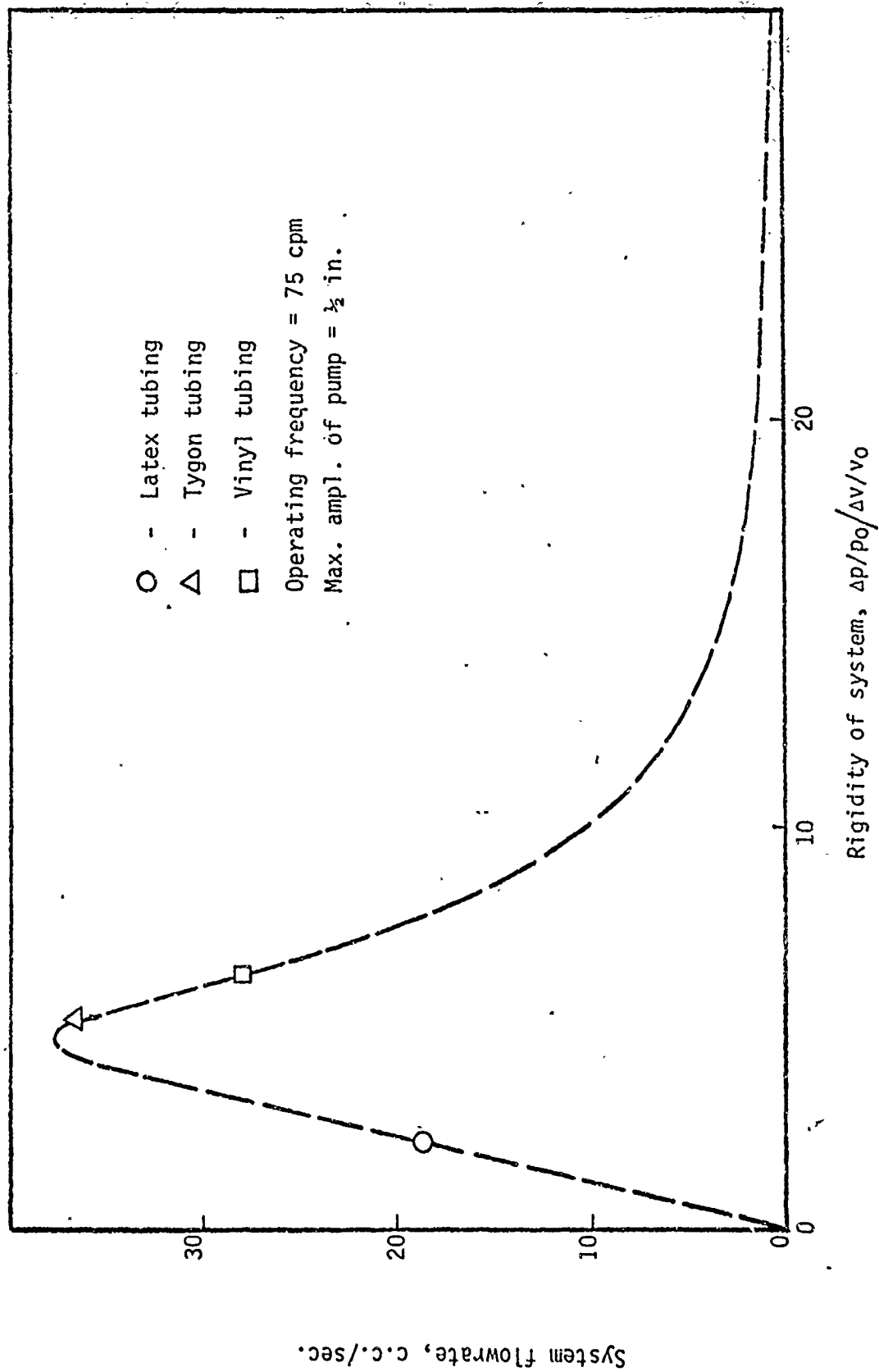


Fig. 12 Variation of flowrate in system with rigidity of tubes

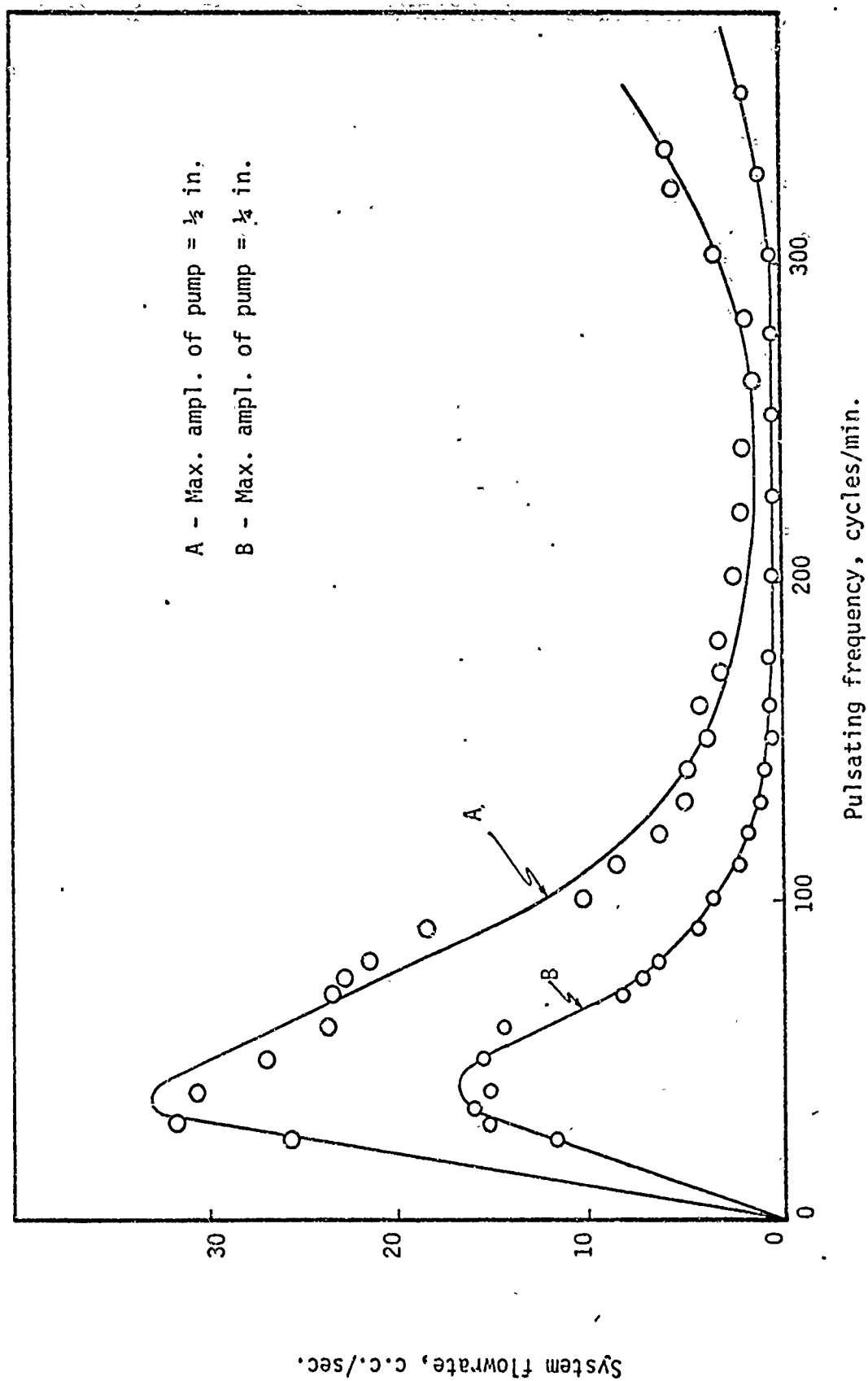


Fig. 13 Variation of flowrate with pulsating frequency for system with steel tubing and open reservoir

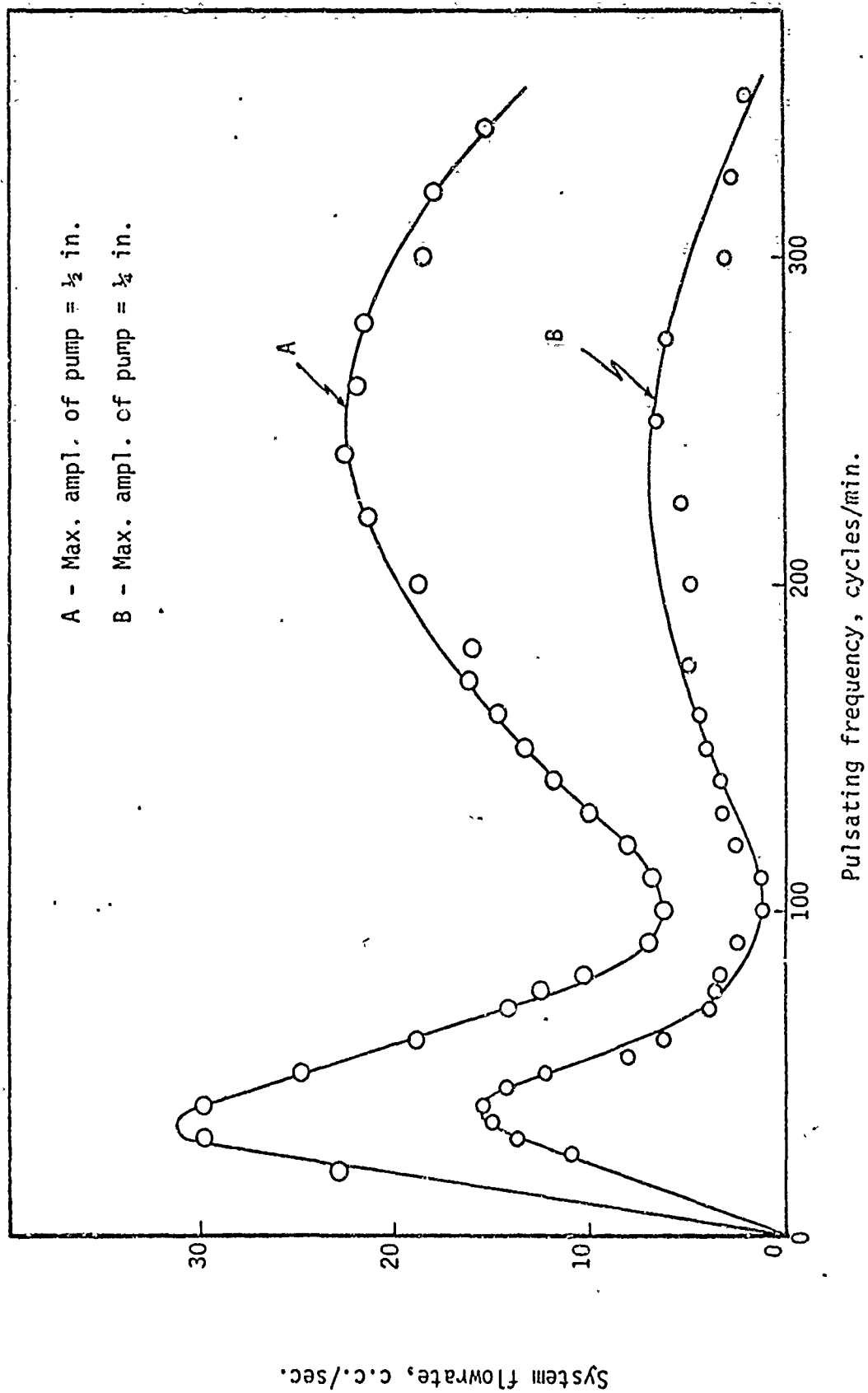


Fig. 14 Variation of flowrate with pulsating frequency for system with vinyl tubing and open reservoir

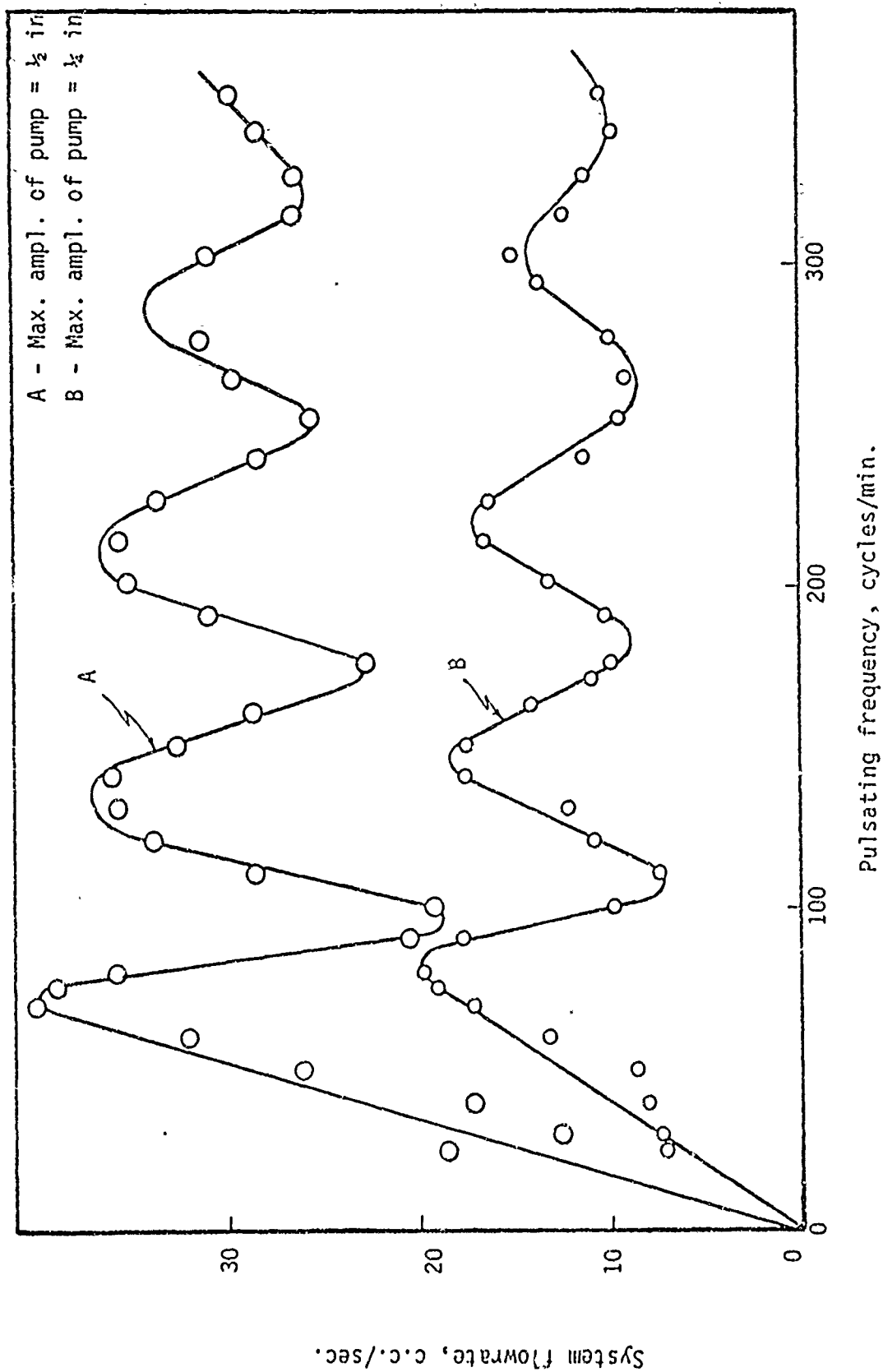


Fig. 15 Variation of flowrate with pulsating frequency for system with latex tubing and open reservoir

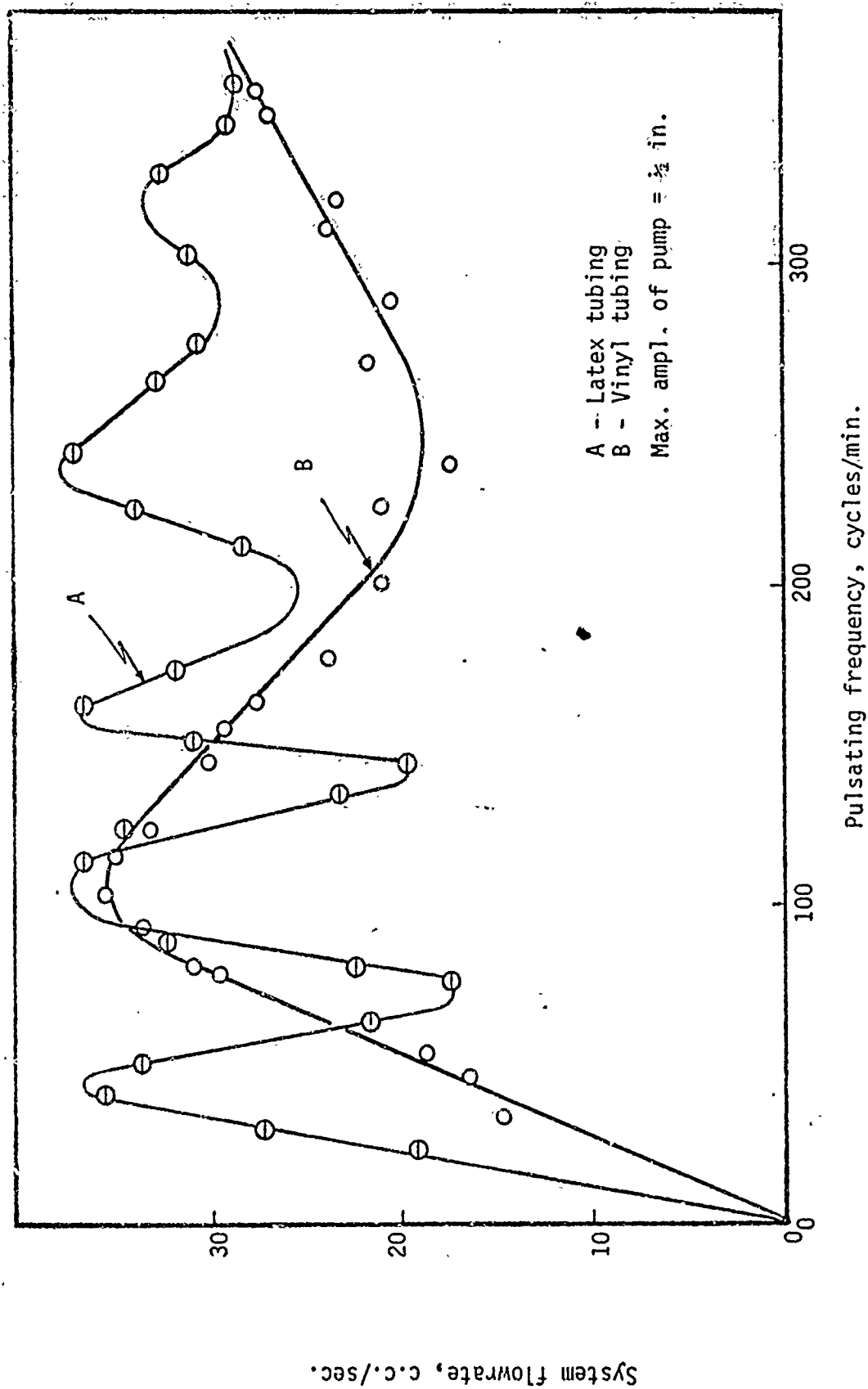


Fig. 16 Variation of flowrate with pulsating frequency for closed system with latex and vinyl tubing